The TumorCNC: Development and Evaluation of a First-Prototype Automated Tumor Resection Device

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Thesis submitted in partial fulfillment of the requirements for the degree of Master of Science in the Department of Mechanical Engineering and Materials Science in the Graduate School of Duke University
2016
ABSTRACT

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Abstract

As technology advances the state of medical imaging, the capabilities of surgical tooling have remained stagnant. This has contributed to a rift between a surgeon’s ability to perceive and their ability to act at small scales. At this point in the evolution of surgical tooling, some level of decision making must be yielded to robotic control. This thesis describes the development and provides an evaluation of a first-prototype device for automated tumor removal. Specifically, the device combines a unique implementation of a three-dimensional scanner with a steerable cutting laser, enabling both sensing and cutting in a platform that can generate 3D images of relatively smooth surfaces to a precision beyond the ability of a human surgeon to act. This device will be used as a research platform to answer the important questions currently standing in the way of bringing automation into the operating room. This work outlines the foundational development of a device that could provide a significant improvement to patient outcomes and reduce operating costs by a magnitude not yet demonstrated in the field of surgical robotics.
For those in the Brain Tool Lab

May the TumorCNC live on
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Motivation and Project Foundation

An estimated 78,000 Americans will receive the news they have a malignant or benign brain tumor this year, and around 45,000 brain tumor surgeries are likely to be performed in the USA alone [1]. At least 33,000 of the patients diagnosed with some form of brain tumor are left to alternative methods of treatment. It is likely that a large percentage of these tumors were deemed inoperable because they are beyond the ability of the current surgical state-of-the-art to remove. For those 45,000 operations that do occur, they are accompanied by documented complication rates between 9 and 40% [2]. This thesis takes a first step in advancing the surgical state-of-the-art through the introduction of a semi-automated surgical robot for tumor resection. Through this effort, both the high rates of complication as well as the number of tumors deemed inoperable can be reduced.

Medical imaging has rapidly matured over the last few decades, but the development of surgical instrumentation has stagnated. These high-resolution imaging modalities assist in tumor identification, but the physical methods used for surgical
removal of diseased tissues have remained fundamentally unchanged since the turn of the mid-20th century. This rift between a surgeon’s ability to perceive minute detail and their capacity to deliver accurate surgical action translates to a direct need not currently fulfilled within medicine. Physicians require a system that can perform tissue manipulation beyond the sensitivity of the human hand. The objective of the Brain Tool Laboratory is to reimagine surgical tooling by focusing on the fundamental goals of precise and efficient tissue removal and tissue preservation.

Even with the advanced level of robotic assistance currently available in the operating room (OR), a surgeon cannot perform at the desired level of precision while constrained by a timeframe so heavily weighted by cost. Some of the basic surgical decisions must be yielded to the assisting robot, thereby giving it a higher level of automation than previously demonstrated in commercially-available surgical robots. The objective of this long-term research program is to design, prototype, and test a semi-automated tumor resection device called the TumorCNC. Informed by multiple medical imaging modalities working in concert to direct the movement of a surgical cutting laser, it will allow researchers in the Brain Tool Laboratory to further explore the introduction of automation into the surgical workflow. Ultimately, this research will lead to a system that provides great improvement in patient outcomes as well as a net reduction in procedure costs as a result of less time spent in the OR.

The TumorCNC removes a tumor in passes, similar to the method in which a CNC milling machine cuts away material layer by layer. After removing each layer, the device interrogates the surgical site with a number of imaging modalities. The current embodiment involves scanning the site with a laser-distance meter which provides a 3D surface map of the zone of interest. Future imaging modalities could include Magnetic
Resonance Imaging (MRI), Ultrasound, Optical Coherence Tomography (OCT), Spectroscopy, and Fluorescence Imaging. The device compares the image data acquired at each layer to the original tumor geometry. It identifies any residual tumor and calculates future cutting paths. This cycle continues under the supervision of the surgeon until the tumor is adequately removed.

One of the principal concerns in the development of the TumorCNC is its ability to resect tissue with great accuracy. By performing laser steering at a higher degree of precision than its human counterparts, the TumorCNC can automate those parts of a surgery that require very little decision making but high physical accuracy – something not previously realized in the operating room (OR). This outcome can only be achieved by focusing on calibration through the extent of the design cycle and accuracy throughout fabrication. Belonging to the Department of Mechanical Engineering, the specific objective of this thesis is to provide a comprehensive introduction to the project, outline the design choices made for the first prototype, and evaluate the physical accuracy of the device. The goal is to prove that a TumorCNC can be developed that demonstrates laser steering capabilities of higher precision than a human surgeon.

The organization of this thesis roughly follows the structure of the design process. It begins with a detailed introduction to the device including an overview of the project architecture, an investigation of market pressures within the current regulatory environment, and an evaluation of related work. The detailed device design is then presented in the context of alleviating the principal concern of steering accuracy. Finally, the calibration procedures are described along with a full system evaluation followed by the discussion and future work.
1.1 Systems-Level Approach

Aside from the neurosurgical principles involved in this project, the robotics effort requires input from a variety of fields including mechanical engineering, computer science, and electrical engineering. Correspondingly, a systems-level approach is critical to the success of the device and its future adoption in the operating room. As such, provision of a high-level project overview first requires identifying stakeholders, outlining a concept of operations (CONOPS), and determining key metrics with which to measure success.

1.1.1 Stakeholder Analysis

Development of a successful device requires the approval of four key stakeholders: hospital administration, physicians, patients, and – within the United States – the Food and Drug Administration (FDA). Three attributes hold importance to these stakeholders: cost-effectiveness, patient outcomes, and the effort imparted by the surgical team. Specific attributes hold more weight with some stakeholders than with others. This distribution is shown in Figure 1.1 where each of the three shaded regions represents a specific attribute, and the stakeholders are placed in those regions in which they hold the most importance. Being able to demonstrate improved patient outcomes is arguably the most important attribute to exhibit, as three of the four stakeholders hold interest in that region. In general, for full endorsement by all stakeholders, the TumorCNC should reduce the net cost to the hospital, provide an overall improvement in patient outcomes, and reduce surgical effort.

The lines between these regions are far from cut-and-dry, but the regions in which stakeholders are presented represent their primary concern. For example, hospital administrators absolutely care for patient outcomes, but their job is to ensure the success
of the hospital. Consequently, I would argue that the attention required of them to keep a balanced budget and maintain the contentedness of their employees supersedes their concern for patient well-being.

![Figure 1.1: Key Stakeholders Organized by Device Attribute](image)

**1.1.2 Concept of Operations**

Creating a CONOPS involves imagining exactly how a system will be used in the environment in which it is designed to be deployed. Examining the CONOPS for the TumorCNC can help identify ways in which the device might fail to deliver on the attributes presented in Figure 1.1. Shown in Figure 1.2, the CONOPS begins with the preoperative workflow as images of the surgical site are captured. In a neurosurgical tumor resection, these generally take the form of MRI or CT image stacks, and the surgeon studies the images while planning the operation. In the TumorCNC workflow, the surgeon first segments these images into a 3D model of the surgical site. The process required to
convert a stack of 2D images into a 3D model has been published on previously and is outside the scope of this thesis [3]–[6].

After segmentation, the surgeon identifies tissue to be removed and plans a surgical trajectory through which the TumorCNC will operate. Both identifying the cancerous tissue as well as planning access to the surgical site are required in a standard tumor resection. The difference in this method is that the surgeon must also communicate this information to TumorCNC – done via transfer of a 3D model of the surgical site to the device.

![Figure 1.2: Concept of Operations](image)

Patient outcomes should not be affected by changes to the preoperative planning stages. Additional steps could lead to more time dedicated to planning by the surgical team which would correspond to an increase in cost and preoperative surgical effort. Consequently, development of the software used to assist in the planning and model generation should focus on usability thereby requiring minimal input from the surgeon.
Following the preoperative workflow, the patient is brought to the operating room, anesthetized, and the surgical team accesses the surgical site in similar fashion to a standard procedure. At this point, the TumorCNC is oriented in a position through which it can operate along the previously-planned surgical trajectory. This location is found through use of a common neurosurgical navigation system such as the Kick Navigation System (Brainlab AG, Munich, Germany).

Once in place, the TumorCNC begins the resection procedure also depicted by the cycle in Figure 1.2. The cycle starts with an interrogation of the surgical site by at least one imaging modality. The received data is processed and converted into an intraoperative 3D model of the surgical site. This model is aligned with the preoperative model through anatomical landmarks either identified by the computer and approved by the surgeon, or identified directly by the surgeon. This alignment is critical to identifying tissue movement, or ‘brain shift’ between cuts. Brain shift is largest during the beginning of the procedure as the brain decompresses from a loss of Cerebrospinal Fluid (CSF), but can continue throughout the procedure [7]. Consequently, after the first calibration, the amount of input required by the surgeon during alignment will likely be reduced as the anatomical landmarks will remain closer to their original position during previous cuts.

After alignment, the tumor volumes are directly subtracted and any residual tissue requiring removal is identified. The TumorCNC composes a cutting path based on a variety of factors including but not limited to the estimated ablation efficiency of the tissue (generated from knowledge of the previous cut), proximity to critical neurologic structures, the amount of tissue requiring removal, the size of the laser spot size, and the range of laser power available. After executing a cut, the entire cycle repeats continuously until the amount of residual tissue is brought to zero or the surgeon aborts the procedure.
In order to accommodate the TumorCNC, the standard surgical workflow for a tumor resection is modified most heavily during the operation itself. In order to maintain cost-effectiveness, the duration of the surgery should not be increased – every minute of robot-assisted operating time can bill over $130 [8]. Patient outcomes can be improved either through reducing complications or decreasing the possibility of recurrence. An increase in precision as compared to a standard resection is the most likely means through which surgical outcomes would improve, as this would theoretically allow for an increased extent of total resection. Finally, the means by which the surgery is performed is very likely to reduce surgeon fatigue. The surgeon will be required to monitor the resection, but the amount of physical activity will be reduced from a standard operation. This reduction in surgical tasking could also increase the number of operating run simultaneously, currently a very demanding task for the attending neurosurgeon.

1.1.3 Design Metrics

Two metrics are directly linked to developing hardware that can demonstrate the three attributes; these are time and precision. For development of this first design, a superficial, small, non-infiltrative, and well-circumscribed tumor as represented by a cortically-based brain metastasis has been chosen as the model case. Combining the target case with the time and precision metrics yields the following two device-level requirements that must be met by any prototype:
The TumorCNC will physically be able to remove the entire mass of a tumor in less than the amount of time a standard resection takes.

The TumorCNC will be able to steer a cutting laser with equal or better precision than the average surgeon manually steering a laser fiber.

It has been previously demonstrated that surgeons have an average precision of 289 microns while traversing a trajectory with surgical tooling, and this is the accuracy benchmark to which the TumorCNC will be evaluated [9]. Unfortunately, benchmarking the device to the time limit of a standard resection requires a barrage of laser-tissue interaction tests which could not be performed in the timeframe of this work. Both speed and precision were considered when developing the first prototype, but only precision will be evaluated.

1.2 Market Analysis

Conducting a market analysis is one method of predicting the development and release of new technology. An accurate prediction of market trends can help physicians to understand what new tools will be released and engineering teams to know what forms of medical device development will be most successful. Post-conceptualization and pre-development, it was necessary to place the TumorCNC in context with the demands of the market.

1.2.1 Financial Overview

From a financial standpoint, the BCC research firm estimated the global surgical robot market to be just under $2 billion in 2014. They predict this number to grow to $2.86 billion by 2020 indicating a compound annual growth rate (CAGR) of 6.4% [10]. Although
a handful of research firms were reviewed, this number is presented because it provides the most conservative growth figure [10]–[13].

More specifically, the predicted CAGR for robotics in neurosurgical applications was estimated to be over 13% [11]. It should be noted that besides surgical robots, this figure includes revenue contributions from surgical navigation systems, surgical simulators, and intelligent operating rooms. At the time of prediction in 2013, the robotics market was worth about $284 million. Due to the immaturity of the neurosurgical robot market, it’s likely that the majority of the systems contributing revenue to the market were surgical navigation systems. Even so, the CAGR for the surgical robot field (6.4%) and the CAGR of the neurosurgical robotics field (13.1%) both indicate significant growth; therefore, it’s highly likely that the neurosurgical robot market will maintain a high rate of expansion.

1.2.2 Regulatory Environment

Contrary to these statistics, the FDA recently released a statement suggesting that no surgical robots are currently operating in the United States [14]. The contradiction lies in a matter of definition. Most robots currently assisting in the operating room can be considered ‘master-slave’ systems – where the surgeon directly drives each movement of the robot through use of a controller. The most common example of this is the da Vinci robot by Intuitive Surgical (Sunnyvale, CA). Alternatively, the FDA does not consider these to be ‘robots’ but defines a surgical robot as having some level of autonomy.

Every surgical robot lies on a spectrum where the most basic are considered to be ‘master-slave’, those that can perform pre-planned actions are considered to have some level of automation, and those that perform actions beyond what was directly planned are considered to be autonomous. The TumorCNC is considered an ‘image-guided’ surgical
robot and would likely be seen as pushing the bounds of automation into the autonomous realm.

Stringent regulation by the FDA is often cited as a counter to market growth within the medical device community. Additionally, it is likely that the lengthy assurance process required by the FDA is a primary factor contributing to a very top-heavy surgical robot market [15]. It is estimated that Intuitive Surgical makes up almost 2/3 of the entire surgical robot market with the majority of the remaining revenue shared between only a handful of other companies [11]. It seems natural that more complex systems - such as those put forth within the surgical robot industry - would require a lengthy approval processes by the FDA. Unfortunately, the corresponding monetary resources required can limit small companies from bringing progressive systems to market. If the FDA did choose to classify the TumorCNC as autonomous, it could make the regulatory pathway extremely costly. This is something to keep in mind when considering the level of autonomy to be exhibited by the device.

1.2.3 **FDA-Approved Neurosurgical Robots**

As a consequence of regulation, only a handful of robots with neurosurgical applications are currently FDA approved and sold within the United States. The majority can be characterized as frameless stereotaxic robots including Renishaw’s Neuromate (Hoffman Estates, IL), Medtech’s ROSA (Syracuse, NY), Mazor Robotics’ Renaissance (Orlando, FL), and IMRIS’s SYMBIS (Deerfield Imaging, Minnetonka, MN) [16]–[19]. Although some of these can be considered image-guided robots, they are still characterized as automated as opposed to autonomous. Each of these robots currently fills a very niche application, but as technology improves and these systems become more refined, physicians
can expect to see an increase in their use within the operating room. The only robot attempting to fill the larger void present in neuro-microsurgical robots is IMRIS’s SYMBIS. Unfortunately, the SYMBIS does not attempt to introduce any automation into the operating room and is currently only FDA approved for performing biopsies [19]. It was developed to be an MRI-compatible ‘master-slave’ robot very similar in capability to the da Vinci.

Not only is the SYMBIS a ‘master-slave’ robot with a very similar surgical approach to the da Vinci, but the estimated price tag also bears a striking resemblance to the costly robot. IMRIS has claimed that the SYMBIS will be sold for around $2 million a unit with an annual service package of $200,000 and a tool cost per procedure of about $1,200 [20]. Following in the footsteps of the Da Vinci, it is likely that the release of the SYMBIS would be met by surgeons and patients with excitement and a corresponding flurry of sales. Similarly, it is unlikely that the bulk of the clinical studies and the health technology assessment will be completed for another decade afterwards. The possibility of competitors entering the market after the adoption of high-cost surgical robots like the Da Vinci or the SYMBIS is low [15]. The vast majority of hospitals have difficulty justifying the value of a single multi-million-dollar robot – to be able to raise enough capital to purchase a second would be a rare case, especially for one with such a niche application. Consequently, robots entering the market after the SYMBIS should attempt to do so with a manageable price tag. This consideration drove many of the design choices for the TumorCNC, as one of the primary marketable attributes of the device is its ability to lower the cost of performing a complex, and resource-intensive surgery.
1.2.4 **Hospital-Level Financials**

Although the market for neurosurgical robots is growing, the overall attitude of the healthcare industry has begun to change from unabated spending to a focus on value and cost cutting. This issue emerges as a prominent theme in all of the market analyses reviewed [11], [15]. Analyzing the acquisition chain within the cost-cutting environment for neurosurgical robots is important in predicting the kinds of technology that will be in demand by hospitals in the near future.

In general, hospitals receive payment for services from private insurance companies and from Medicare. These payments are typically in the form of fixed reimbursements depending on the surgical procedure performed [21]. This financial relationship strongly effects the hospital’s bottom line which ultimately influences their decision-making process on surgical-robot investment. Specifically, hospitals are reimbursed approximately the same amount whether a surgeon performs a procedure using a multimillion-dollar surgical robot or performs the same procedure manually. Multiple studies have shown that procedures using surgical robots such Intuitive Surgical’s Da Vinci cost the hospital significantly more [21]–[24]. This extra cost must be offset by either attracting more patients or by performing procedures the surgeons would not be able to perform without the robot.

It has been shown that nearly half of all hospitals advertise for robotic surgery on their websites, and the vast majority claim clinical superiority [25], [26]. If a large part of the calculus behind purchasing a surgical robot is based on attracting patients through marketing, it would be critical for companies attempting to release high-cost surgical robots to launch marketing campaigns that not only target hospitals and surgeons, but that also stir up demand for their product amongst patients. Other alternatives for medical
device designers is to focus on developing tools that will allow hospitals to cut costs by saving time or by reducing the amount of staffing required per procedure. Ideally, the TumorCNC will accomplish this through an increase in automation which would correspond to a reduction in the amount of human participation during a tumor resection. This could potentially reduce the amount of time in the OR and minimize the number of surgical support staff.

Investigating the drivers behind and the resistance to market maturation can be useful for predicting trends in emergent technology. Several forms of resistance have been identified including cost cutting, competition, and federal regulation. There is still a strong push for innovation as doctors seek improvement in patient outcomes. Furthermore, physicians preference items continue to make up around 30% of hospital budgets and surgeons will continue to choose tooling that is comfortable and easy to use regardless of administrator demands to cut cost [27]. The primary drivers behind market expansion appear very similar to those 5 years ago [11], [15].

1.3 Previous Work

Distanced from the market pressures driving commercial robotics, academics have been exploring the role of automation in the operating room for decades. That said, only recently have advances in robotics progressed the field enough to allow automated surgical robots to push the bounds set by their human counterparts. An analysis of previously published related work allows designers to predict and address the largest challenges in the field.

Very few devices have been developed which attempt to introduce automation through a dependency on intraoperative imaging. A handful of image-guided robots have been developed, but these generally operate based on paths planned from preoperative
images. With their own unique challenges, each of the two devices introduced below utilize intraoperative imaging to provide a sensor stream that ensures the accuracy of the system.

1.3.1 *A First Image-Guided Surgical Robot for Tumor Resection*

Dr. Patrick Kelly of the Mayo Clinic was the first to utilize what could truly be called an image-guided surgical robot in the OR. Described in a series of articles published between the late 1970s and the late 1980s, his group set the stage for the future of surgical automation by resecting 83 neoplasms under full automation [28], [29]. With the modest computational resources present during the 1980s, they were able to show that there was no decrement to patient outcomes when performing a computer-assisted resection for those deep-seated tumors that are difficult for a surgeon to localize.

The robot developed by their team is very similar in concept to the TumorCNC. The resection tool was a Carbon Dioxide (CO₂) laser that vaporized the tumor in layers [29]. In similar form, their surgical plan was based on preoperative images (CT), but they also employed an X-Ray to intraoperatively monitor the procedure. Prior to beginning resection, the surgeon placed bearings at known depths within the tumor via stereotaxic cannula as a form of landmark. These bearings allowed the surgeons to adjust their imaging to account for brain shift and measure cut depth. Depth was also monitored via calibration marks located on the stereotaxic retractor.

Their team was met with a variety of challenges that likely prevented adoption of the device within the neurosurgical community. What seems to be the most difficult challenge affecting patient outcomes was the accuracy of the laser. From a baseline hardware standpoint, their stereotaxic frame was able to achieve an accuracy of approximately 800µm on average [30]. Combining this with a 1mm segmentation resolution
when planning a resection volume yielded an error of approximately 1.3mm – more than four times the average precision of a surgeon [29]. These numbers do not account for the fact that the placement of the bearings also depended on the stereotaxic frame, and the depth monitoring was based on recovery of these 1mm-diameter bearings and a 2D X-Ray taken from the side of the surgical site.

From an efficiency standpoint, the surgical workflow was fairly cumbersome in comparison to a standard resection. Although a lot of the workflow could have likely been streamlined given widespread use, there were a handful of steps based on technical requirements that could not be avoided. These problems were rooted in the methodology used to monitor depth. Without access to the intraoperative imaging technologies becoming available now, their team was dependent on an X-Ray to give truth data on depth. Embedding bearings for landmarks, requiring manual monitoring of the X-Ray, and managing a bulky stereotaxic frame would have caused an unavoidable increase in procedure length and cost.

1.3.2 Automated Tumor Resection – A Modern Approach

What could be considered the most modern embodiment of Dr. Kelly’s work is a system developed by a group in Tokyo also to be used for brain tumor resection [31]. Presented in a series of publications between the mid-2000s and 2013, they linked intraoperative MRI (IoMRI) and 5-aminolevulinic acid (ALA) fluorescence to a laser steering device as a form of intraoperative guidance. This intraoperative guidance acts as a replacement for the bearings planted by Dr. Kelly’s team which were used to indicate the intraoperative extent of resection and brain shift.
Both IoMRI and 5-ALA fluorescence have been shown to increase the extent of resection and consequently, improve patient outcomes [32]. The unfortunate reality is that IoMRI and 5-ALA take far too much OR time to be used cost-effectively during an automated resection. The system implemented by Liao’s team took ‘several minutes’ to scan a 1cm square of tissue with fluorescence, and a review of the use of IoMRI found that a single scan would add on average of 45 to 60 minutes to a case [33]. Performing a scan with either (or both) of these modalities at every cut would prohibitively lengthen time in the operating room.

Like Kelly, Liao’s team may have built something well before its time. Advances in IoMRI and fluorescence could increase feasibility of the device. As it was left, an evaluation of consecutive cuts during a full volumetric resection in either phantoms or animal models was not reported, nor was a reliable, system-level accuracy assessment. At this point, it’s important to field a device that clearly brings a precision greater than that of a surgeon while streamlining the surgical workflow and reducing operating time; unfortunately, the Liao device offers none of these benefits to date.
Device Design

Transforming the conceptual surgical workflow outlined in Chapter 1 into a working device is a complex endeavor. This chapter details the design choices made to assemble the first prototype, and it is organized primarily by the primary components that make up the system. These include a cutting laser, a triangulation-based distance meter used to generate a 3D model of the surgical site, and a galvanometer used to steer both of these components. The final sections detail the optical and electrical integration and the software developed to drive the prototype.

The first prototype was assembled on a 2-foot by 3-foot optical breadboard in order to simplify construction and accommodate any unforeseen changes. Figure 2.1 shows an image of the device with the main components identified. The enclosure surrounding the breadboard consists of ¼-inch acrylic panels covered in blackened aluminum; its purpose is to stop any stray laser beams at 10.6 µm from leaving the breadboard (designed and constructed by assisting graduate student David Britton). The beam paths are shown in
Figure 2.2, where the cone-shaped area in yellow indicates the range of the triangulation sensor – the surgical site can be located anywhere in that area. Clearly, in its current embodiment, the TumorCNC will not be introduced into the OR. Making this transition would require a significant size reduction and careful consideration in packaging such that it can be fully draped and sterilized. The components in the green box of Figure 2.2 make up the heart of the device. Building a prototype for the operating room would involve repackaging these components into an enclosure to be used near the patient, while the remainder of the optical and electrical components would be located in a rack to the side of the bed with the lasers transferred by fiber optic.

**Figure 2.1: TumorCNC Prototype – As Built**
2.1 The Cutting Laser

The cutting laser is arguably the most important component of the system as it accomplishes the primary goal of removing tissue. Lasers are generally characterized by four distinct properties, the wavelength, power output, pulse length, and spot size. Variations in any of these factors change how the laser pulse interacts with the tissue, but the most dramatic changes occur through a variation in wavelength.

The absorbance of light in tissue is dependent on the wavelength of the light as well as the presence and distribution of light-absorbing chromophores such as water, hemoglobin, or collagen [34]. Lower absorbance allows light to penetrate further into the tissue corresponding to a wider distribution of heat, while higher absorbance corresponds
to a shorter penetration depth and a more confined temperature distribution. These distributions drive the ablation properties of that specific laser-tissue combination.

2.1.1 Neurosurgical Lasers

The search for the primary cutting laser for the TumorCNC was constrained to lasers that have been or are currently used for neurosurgical tissue removal. This constraint allows our team to leverage the large body of literature that currently exists describing the laser-tissue interaction between these specific neurosurgical lasers and brain tissue. Constraining the search to lasers already FDA approved for neurosurgical use will also allow for easier acceptance and greater recognition from the neurosurgical community.

Within the neurosurgical realm, two varieties of surgical laser are commonly employed for tissue removal, the neodymium-doped yttrium aluminum garnet (Nd:YAG) laser and the carbon dioxide (CO₂) laser [35]. Between the two lasers, the $\lambda = 10.6 \ \mu$m wavelength of the CO₂ (as compared to $\lambda = 1.06 \ \mu$m for the Nd:YAG) experiences a much shorter penetration depth in brain tissue due to its high absorbance by water. This characteristic allows the CO₂ much greater cutting precision and far less thermal spread than the Nd:YAG. Unfortunately, minimizing this thermal spread prevents the CO₂ laser from coagulating blood vessels as effectively as the Nd:YAG. This tradeoff between precision and the ability to achieve hemostasis means the CO₂ laser is generally only able to coagulate veins and arteries with diameters less than 0.5mm [36].

Because precision is one of the driving factors behind the development of the TumorCNC, the CO₂ laser was selected as the primary cutting laser. The specific characteristics were modeled after one of the most popular FDA-approved CO₂ lasers within neurosurgery: The OmniGuide FELS-25A (OmniGuide Surgical, Lexington, MA).
This specific laser gained popularity due to OmniGuide’s patented hollow-core fiber optic – currently the only FDA-approved method to transmit $\lambda = 10.6 \ \mu m$ light through a fiber [37].

The stereotaxic approach employed by the TumorCNC requires that the cutting laser is focused down to a well-known diameter at the surgical field. Using a fiber-launched laser would require a number of optics to be installed near the face of the fiber to collimate and focus the expanding Gaussian beam onto the surgical target. Consequently, the free-space Synrad 48-1SAL (Mukilteo, WA) was installed as a surrogate to the fiber-launched OmniGuide on the first prototype of the TumorCNC. The Synrad can replicate the continuous mode and many of the pulsed modes of the OmniGuide at a far more economical price tag ($350$ used on eBay as opposed to $15,000$ from OmniGuide).

Other lasers that could be effective in this application have been identified outside of the neurosurgical realm. The reason to select one of those over the CO$_2$ laser would be to provide cleaner and more predictable ablation. These lasers are discussed in the final chapter of this thesis.

2.1.2 Laser-Beam Path

Integrating the cutting laser into the system requires injection of a Gaussian beam into the steering galvanometer with a well-known focal point, spot size, and depth of field (DOF). Determination of these beam characteristics will allow the control software to generate accurate and concise cutting paths to the benchmark set in the first chapter.

The first stage of the system is the expansion region, where the emitted beam is allowed to expand naturally (at a full-angle divergence of 4 mrad) from a width of 3.5mm to a diameter of approximately 6mm (as shown in Figure 2.3) [38]. The emitted beam is
square in profile, but the expansion region allows this profile to diverge to a somewhat-
circular, near-Gaussian beam after reaching the far field at a distance of approximately
200mm. All mirrors in the laser line are protected silver with a surface flatness better than
100nm, a reflectance greater than 95% at 10.6µm, and a scratch-dig specification of 40-20
[39]. These specifications allow for a balance between economy and performance. Mirrors
with better scratch-dig specifications were not selected because the expansion region is
located before the spatial filter where imperfections in the Gaussian profile can be
eliminated; consequently, the extra cost is not merited.

\begin{figure}
\centering
\includegraphics[width=\textwidth]{spatial_filter_diagram.png}
\caption{Laser-Beam Path}
\end{figure}

The spatial filter consists of a Keplerian telescope with a pinhole placed at the
focal point of the first lens (see the detail view in Figure 2.3). The system was designed to
accept one of three different lenses on the objective side of the telescope. Placing a lens
with a longer focal point allows the beam to expand further, permitting for a smaller spot
size after the final focusing element. In general, a larger beam incident on the final focusing
lens corresponds to a smaller beam waist at the focal point. Consequently, the orange lens
in Figure 2.3 corresponds to the largest spot size at the surgical field while the blue lens corresponds to the smallest spot size. For simplicity, all optional objective lenses (the colored lenses) are shown in the figure although only one will be installed during operation. The corresponding difference in beam diameter is shown as it enters and exits the final focusing lens for all three configurations. Table 2.1 illustrates the corresponding range of possible spot sizes at the targeted tissue as calculated theoretically [40]. The mode quality factor \( M^2 \) is less than 1.2 as specified by the laser manufacturer [38]. Consequently, the spot size can be bounded between \( M^2 = 1.2 \) and \( M^2 = 1 \) and the average rounded output is given as the reference spot size.

**Table 2.1: Calculated Laser Spot Size**

<table>
<thead>
<tr>
<th>Telescope Objective Lens Focal Length [mm]</th>
<th>20</th>
<th>40</th>
<th>75</th>
</tr>
</thead>
<tbody>
<tr>
<td>Target ( \frac{1}{f} )</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Spot Size [( \mu m )]</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>( D = \frac{2M^2\lambda f}{\pi r_i} )</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>( M^2 = 1.2 )</td>
<td>1080</td>
<td>540</td>
<td>288</td>
</tr>
<tr>
<td>( M^2 = 1 )</td>
<td>900</td>
<td>450</td>
<td>240</td>
</tr>
<tr>
<td>Reference Spot Size [mm]</td>
<td>1</td>
<td>( \frac{1}{2} )</td>
<td>( \frac{1}{4} )</td>
</tr>
</tbody>
</table>

The lenses used to change spot size are replaced manually based on the resolution required in the specific procedure. Contrary to this approach, the device constructed by Liao et al. implemented an autofocus mechanism to ensure the focal point of the cutting laser is incident on the surface of the target tissue [31]. In order to simplify the design of the first prototype of the TumorCNC, the autofocus was omitted but should be considered for future iterations. In its stead, the final focusing element was selected to have a long focal length, allowing for a wide DOF characterized by a slowly changing spot size as a given target is moved outside of the focus. The chosen focal length of 200mm also allows the smallest spot size to occur near the most accurate regime of the triangulation device (described in the following section).
All lenses in the laser line are plano-convex and spherical, ground from zinc selenide (ZnSe) with an anti-reflective (AR) coating optimized for $\lambda = 10.6 \, \mu m$. ZnSe was chosen because it passes 10.6 microns with nearly 100% transmission when paired with the proper AR coating [41]. Spherical lenses were elected over aspherical in an effort to reduce cost; therefore, the optics were sized to prevent incident light from hitting more than 50% of the lens to reduce the amount of spherical aberration commonly introduced from the edge of a spherical element.

2.2 The Distance Meter

Similar to the ALA fluorescence and IoMRI used by Liao’s device and the x-ray on Kelly’s device, the distance meter is used to provide intraoperative information to the TumorCNC [29], [31]. Its intent is to serve as a 3D scanner; by combining it with the scanning galvanometer, it can generate a surface map of the surgical site which allows for the tracking of each laser cut with relation to the preoperative model. Although an imaging modality such as IoMRI would allow for full 3D volume generation, in its current state it requires too large a footprint in the OR and too much operating time to provide a viable alternative. Three-dimensional scanning was deemed to have the lowest price point and minimal intraoperative impact while physically maintaining the ability to produce a model of the surgical site with the required level of precision.

2.2.1 3D Scanning in Medicine

The use of 3D scanning has some precedent in medicine. Its primary applications are in dental and plastic surgery, where the patient’s anatomy may be scanned to assist with planning an operation or designing an implant [42], [43]. Realizing a void in the field of
surgical navigation, a group at Vanderbilt University recently started a company with the goal of marketing a 3D scanner developed for the operating room [44], [45]. This device has also been tested on the cortical surface of the brain ex vivo, setting the precedent for the use of 3D scanning in a neurosurgical application [46]. Unfortunately, within their comparative study of three scanners, none were precise enough to satisfy the requirements of the TumorCNC.

With these studies in mind, identifying a 3D scanner with the necessary precision became a challenge. A variety of scanner types exist in the commercial domain including time of flight, conoscopic holography, laser triangulation, structured light, and stereo-image pairing. Based on the short range necessary from device to surgical sight, time of flight sensors were eliminated. Devices utilizing stereo-image pairing seemed to be unable to offer the appropriate resolution, and structured light sensors tend to have problems with reflections on shiny or wet surfaces. Conoscopic holography and laser triangulation were both found to offer precision values far better than the required benchmark of 289µm, but laser triangulation was selected with the more economical price tag.

2.2.2 Triangulation Sensing

The specific triangulation sensor used on the TumorCNC is a model 200-300 Microtrak 3 (MTI Instruments Inc., Albany, NY) [47]. With a 300mm range, the sensor allows a maximum standoff of approximately 8 inches from the front of the steering galvanometer to the surgical target. The sensor directs a red laser beam at 670nm through the galvanometer and onto the surgical site. The image of the laser on the targeted surface is reflected back through the galvanometer and directed through a lens onto a line of charge-coupled device (CCD) pixels. The image of the spot is incident on different pixels based
on the angle at which it passes through the lens, uniquely identifying the distance of the
target. The spot size of the Gaussian beam is quoted to be 130µm at standoff; however,
the lateral accuracy is generally smaller than the spot size as the image is spread across a
variety of pixels. The operating software thresholds the return of each of the pixels based
on intensity and then takes the centroid of the remaining pixels. A diagram of the
triangulation sensor and its method of operation is shown in Figure 2.4.

\begin{center}
\includegraphics[width=0.5\textwidth]{triangulation_sensor.png}
\end{center}

\textbf{FIGURE 2.4: Triangulation Sensor}

To build a 3D model of the surgical site, the laser emitted by the triangulation sensor is
steered across the surface of the target by the steering galvanometer. As the galvanometer
moves, the triangulation sensor continuously samples at its maximum frequency of 20kHz.
By pairing the instantaneous angle of the galvanometer mirrors with the corresponding
distance sample, a 3D surface map is created.

\section*{2.3 The Steering Galvanometer}

The third component of the system is the steering galvanometer – a mechanism consisting
of two motors with their axes oriented perpendicular, each driving a mirror. These motors
provide independent steering along two different axes. The galvanometer installed in the
TumorCNC (Figure 2.5) is a model 6240H by Cambridge Technology (Bedford, MA).
Silver mirrors were installed of a size allowing for the passage of a 25mm-diameter image. Although the diameter of the cutting laser does not increase beyond one centimeter, large mirrors were required to allow for the reflection of both the triangulation sensor laser as well as its reflected image.

![Steering Galvanometer](image)

**Figure 2.5: Steering Galvanometer**

Along with the drive circuitry introduced in Section 2.5, the galvanometer has the potential to introduce a significant amount of error into the device. A Cambridge Technology galvanometer was selected because their product specifies a tolerable amount of error below the requirements outlined earlier in this thesis. Furthermore, the mirror mounts on the Cambridge Technology galvanometer allow for field replacement of the mirrors if necessary.

2.4 Optical Integration

Bringing each of these components together into a functional system requires a significant effort in alignment and calibration, the bulk of which is discussed in the following chapter. A few other components are also necessary to make this a comprehensive first prototype; a guide laser, a beam combiner, and an optical mount.
2.4.1 The Guide Laser

A common complaint between users of the CO$_2$ laser is that the emitted wavelength is invisible [48]. Without a guide laser, the surgeon doesn’t necessarily know where their cutting tool is pointing. This could be especially unnerving with the TumorCNC as the surgeon is not manually holding the cutting fiber. Consequently, a helium-neon (HeNe) gas laser was incorporated into the design of the device. The specific laser is a Melles Griot 05-LHP-121 at a wavelength of 633nm. It is a 5mW laser with a 3R classification, thereby making it bright enough to be visible under any lighting while preventing the unnecessary regulation introduced by a class 4 laser. A red wavelength was also chosen because beam combiners built for a HeNe CO$_2$ combination are far more common at the red wavelength of 633nm.

The HeNe is mounted parallel to the triangulation sensor but in the same optical plane as the CO$_2$ laser. Its beam is reflected off of two silver mirrors. The first directs the beam towards the triangulation sensor, the second is mounted directly between the emitter and the sensor on the triangulation sensor (labeled as the ‘Guide Mirror’ in Figure 2.6). This mirror aligns the beam to be roughly parallel to and in a vertical plane with the triangulation laser.

2.4.2 The Beam Combiner

With both the triangulation sensor and the HeNe running parallel to each other, it is necessary to merge them with the cutting beam such that all beams enter the galvanometer within the required field-of-view of the mirrors. The beam combiner selected to merge these beams is of ZnSe construction with an AR coating on one side (optimized for 10.6µm) and a reflective coating on the other side (optimized to reflect 633nm). These coatings
allow the optic to pass a 10.6 m wavelength at almost 100% efficiency while providing acceptable reflectance to red light. Similar to the sizing of the galvanometer mirrors, the beam combiner is 1.5 inches in diameter – large enough to reflect both the emitted triangulation beam as well as its image.

2.4.3 Mount Construction

Careful consideration of how each of these components are mounted prior to assembly can greatly simplify alignment and calibration. Consequently, a single mount was designed to interface with the triangulation sensor, the galvanometer, the beam combiner, and the spatial filter of the cutting laser. This mount was machined from a single piece of aluminum, putting each component in a known position within the bounds of tolerance quoted by the machinist. Each surface requiring close machining tolerances is labeled in Figure 2.6.

![Figure 2.6: Optical Integration Mount](image)
In order to further ensure tight tolerances, the mount was designed such that all critical surfaces could be machined from a single orientation in the CNC mill. By this methodology, it was unnecessary for the machinist to chuck the part multiple times, meaning those critical surfaces are within a tolerance of .0005 inches as specified by the manufacturer of the CNC mill (Bridgeport – Hardinge Inc., Elmira, NY). Minimizing these tolerances is critical to the calibration process detailed in the following chapter.

2.5 Electronics

The final section of this chapter describes the electronics and drive software developed to operate the first prototype of the TumorCNC. Two figures provide sufficient detail to describe the design. Figure 2.7 shows the prototyped control panel and identifies each of the main components while Figure 2.8 shows the principal electrical connections. The dotted green box outlines those components mounted onto the control panel.
The interface between the hardware and the computer consists of an Arduino Micro and a National Instruments USB-6211 Data Acquisition Card (DAQ). The Arduino provides a link to each of the lasers while the DAQ controls steering and measurement.

### 2.5.1 Laser Control

Three lasers are controlled through the Arduino depicted in Figure 2.8; these are the CO$_2$ cutting laser, the HeNe guide laser, and the diode laser embedded in the triangulation...
sensor. Both the HeNe guide laser and the triangulation laser are controlled through solid-state relays connected to digital output pins on the Arduino. For the HeNe, the relay directly controls access to the 120VAC supply, while on the triangulation sensor it grounds a control pin which activates the laser diode. The computer needs the ability to activate these lasers because during scanning, if both lasers are on, two beams of similar wavelength will be incident on the sample thereby confounding the triangulation sensor. When in cutting mode, the distance meter needs to be turned off in order to prevent surgeon confusion as a result of a secondary red spot misaligned with the cutting laser appearing in the surgical site.

Aside from control of its laser, the other setting available to the consumer on the triangulation sensor is its frequency of data generation. The fastest available rates are 20kHz, 4kHz, and 1kHz where these rates determine the integration time of the receive sensor. With too fast a rate on too dark a surface, none of the pixels available on the sensor will reach the cutoff threshold described under Section 5.2, and the sensor would miss that measurement. Control of these rates is set via a driver supplied by the manufacturer (MTI Instruments, Albany, NY).

For the cutting laser, both the power and pulse width are controlled via PWM through a digital-output pin on the Arduino. The laser requires pulses of a very specific frequency and duration to energize the CO$_2$ gas prior to generating a high-energy pulse – these are referred to as tickle pulses [38]. The required pulse width of this tickle is 1s, thereby setting the maximum clock rate of 1MHz which is implemented via Arduino.

The Arduino Micro used to control the CO$_2$ laser has a hardcoded PWM frequency of 500Hz. Consequently, the timers available on the microprocessor need to be manually set in assembly language to activate certain registers on a timed interrupt (a script
skillfully developed by collaborating graduate student Weston Ross). The use of these timers to provide the tickle pulse puts several limitations on the cutting-pulse duration. In the current embodiment, the shortest pulse available is approximately 250 s with longer pulses generated as multiples of 125 s. These numbers are not hugely limiting as the shortest pulse width of the laser is physically restrained by an approximate rise time of 100 s. An exploration of the impact of laser pulse-width is discussed in the final chapter of this thesis. The power output of the cutting laser is dictated by the duty cycle of the PWM input into the system as a percentage. The correlation between duty cycle and physical power output is presented in Chapter 3 – Calibration and Testing.

2.5.2 Data Generation and Acquisition

All remaining data passed and received from the TumorCNC is done through a USB-6211 Data Acquisition card (National Instruments). This data includes all positional commands sent to the galvanometer and all of the distance measurements received from the triangulation sensor. All communication between the DAQ and the galvanometer is analog; any given voltage corresponds to an angular position of the mirrors. Consequently, the DAQ has two analog output channels – one dedicated to the x-motor on the galvanometer and one dedicated to the y-motor. The motors accept a range of voltages between -10 and 10 Volts, where each volt corresponds to 1 degree of rotation and 0 Volts corresponds to the center position. Note that each degree of mirror rotation corresponds to a 2-degree change in the angle of the output beam because the input and output beam angles are changed by 1 degree.

The galvanometer is also configured to output a variety of status signals as analog voltages ranging from -10 to 10 Volts including the position, positional error, and velocity.
For the first prototype, the DAQ was setup to accept position and error. The scale of these is the same as the output voltage were each Volt corresponds to 1 degree of rotation on the motor. As soon as the galvanometer controller receives an angular position command, the positional error is calculated by subtracting the commanded position from the actual position. Monitoring this signal allows the computer to determine when the galvanometer has reached its desired position. Similar to the galvanometer, the input from the triangulation sensor is also analog. A measured distance lying within the range of the sensor is converted to a 16-bit value which is then converted to an analog voltage between 0 and 10 Volts and is sampled by the DAQ.

Unfortunately, this dependence on analog signals is relatively fragile as it requires high accuracy from the digital to analog/analog to digital converters as well as significant noise isolation within the electronics. All the connections are made with twisted pairs to improve noise rejection and all power supplies were selected with a ripple factor better than that suggested by the manufacturer of the device. Furthermore, all power supplies are connected to chassis ground in order to prevent a floating ground that could offset the analog voltage of a single device. Further impacts on the accuracy of the system are discussed in the following chapter – Calibration and Testing.

2.6 Drive Software

Aside from the code developed to control the Arduino, all of the software used to run the TumorCNC was built in Python 2.7. The operating script primarily calls two classes: one is dedicated to converting galvanometer-space to Euclidian (and vice-versa) and one is used for interfacing with the DAQ. Developing each of these had its challenges – performing the conversions required solving a set of difficult geometric problems, while
interfacing with the DAQ required the use of a convoluted, python-wrapped C driver called PyDAQmx (Pierre Cladé). The goal was to develop a set of tools which will allow the system to operate at the full potential of the hardware. It should also be modular such that when new functionality is required, the base-level DAQ control class and the angular conversion class can simply be imported into the new code base.

2.6.1 Controlling the DAQ

Building the software to control the DAQ starts by examining the end goals of the TumorCNC. There are instances during a procedure where it could be necessary to use either velocity control or position control of the galvanometer mirrors. The only available input to the galvanometer is position, but these analog commands can be updated with a fine enough resolution to mimic velocity control. Consequently, the DAQ has been configured to output in two different modes – continuous (which implies velocity control) and discrete (which implies position control). In both modes, the script is programmed to accept a list of angle commands to send to the \( x \) and \( y \) mirrors. Depending on whether the scan will be used for cutting or for measuring, the pulse length of the laser or the measurement frequency of the triangulation sensor is also required as an input.

When scanning in discrete mode, the DAQ outputs a position and waits a specified amount of time for the galvanometer to settle on the commanded position. Once the settling time has passed, the DAQ samples the triangulation sensor or sends a digital pulse to the Arduino requesting a pulse from the cutting laser depending on whether it is in cutting mode or 3D scanning mode. The DAQ waits for the sample to be fully acquired or for the cutting pulse to complete, then repeats the process. In the current implementation of the discrete mode, the DAQ driver is configured to start and stop the analog output
and analog input tasks at every position of the galvanometer. In future implementations, one of the internal clocks will be used to trigger each sample as some amount of time is lost to starting and stopping the task at each step. Furthermore, instead of waiting for a specified settling time at each command, the DAQ will continuously monitor the position of the galvanometer and trigger the sensor afterwards. In general, the settling time of the galvanometer and the sampling time of the triangulation sensor are both on the order of tens of microseconds, so even without these improvements, the discrete scan is able to gather large numbers of samples in a small amount of time.

Alternative to the discrete scan, the continuous mode scans the surface of the target without pausing to gather data or execute each laser pulse. As described above, each sampling position in discrete mode is accompanied with a set of galvanometer angles. In continuous mode however, the DAQ outputs galvanometer angles at its maximum update rate by adding interpolated data points between the desired sampling angles. The output is a much more continuous scan. The speed of this scan is controlled by changing the sampling rate of the triangulation sensor (slower sampling rate corresponds to a slower scan), or by changing the resolution at which the user wishes to gather samples (finer resolution also corresponds to a slower scan).

Building a 3D surface from triangulation-sensor data requires correlating each distance sample with the angular position of each mirror at the instant that sample was measured. In continuous mode, the angular position is found by averaging all of the position points since the last sample was received. Consequently, the advantage of continuous over discrete is that scans can occur much more quickly, but because each measurement is taken as an average of the positional output, it can be argued that the accuracy is diminished as compared to a discrete scan. In practice, continuous mode would
be used when performing large surface scans and executing bulk resection with high power over large areas. The discrete scan becomes useful near the fringes of a tumor where higher precision is necessary in measurement as well as in cutting. Hypothetically, depending on the amount of tissue to be removed, the laser power can be tuned for every shot in discrete mode.

2.6.2 Stereotaxic Conversion

Stereotaxic is a commonly-used term in neurosurgery that refers to the surgical approach where the patient’s anatomy and the surgical tooling are converted into a single reference frame. This approach can be applied to the TumorCNC, as the patient’s anatomy needs to be easily referenced when cutting, measuring, and working with preoperative models of the surgical site.

The common coordinate frame for the system was chosen to be TumorCNC-centric, where a Euclidian frame is oriented with reference to the galvanometer. Intuitively, with angular commands sent to the galvanometer, it would seem that a spherical frame would be the best choice for a coordinate system. However, because the $x$ and $y$ mirrors are located in series and the cutting laser and triangulation laser do not intersect the mirrors at the same point, a spherical system would have been an extremely difficult choice. Instead, the Euclidian frame was defined such that the $x$-axis is aligned with the axis of the $y$-motor on the galvanometer, and the axis of the galvanometer $x$-motor is located in the $y$-$x$ plane with the $z$ axis pointed toward the target. The coordinate frame is shown for reference in Figure 3.3 in the following chapter.

Unfortunately, a variety of factors make the conversion between angular galvanometer commands and the Euclidian frame very difficult. Due to constraints in
positioning, neither the cutting laser nor the triangulation laser intersect the galvanometer mirrors at their center point, meaning any change in mirror angle also corresponds to a slight translation of the beam. An independent translation occurs as a result of the reflecting faces of the mirrors not being centered on the axes of their respective motors. To make matters worse, the axis of the galvanometer x-motor was designed to be at a 14-degree upward tilt in reference to the base of the device, introducing another rotation when attempting a conversion. Finally, although the cutting laser and triangulation laser were designed to be oriented along specific axes of the galvanometer, the mounting accuracy can never be perfect meaning calibration factors must also be taken into account during angular conversions.

The simplest method of accurately including all of these factors was to generate a ray-tracing model within the conversion script. This model uses vector-based operations to trace each ray through the geometry of the system, tracking the distance traveled and each point at which it intersects a mirror, optic, or target. The first step in creating this model was to build an as-designed model of the TumorCNC in SolidWorks. Models for each component of the system were either imported from the manufacturer website or generated from dimensional drawings. Following the construction of this parametric model, the ray-tracing script was developed in Matlab with each surface represented as a pair of vectors, one for the rotation point of that surface and one for the corresponding unit-normal vector. To ensure accuracy, rays were randomly generated, traced through the model, and compared to the parametric truth-model created in SolidWorks. After a full model validation, the Matlab code was converted to Python and made to accept lists of angles to convert, thus increasing the speed of calculation to the level required by the TumorCNC.
The architecture of the python-implemented script is presented in Figure 2.9; the top describes each of the basic functions included, and the bottom shows how the script converts galvanometer rotation commands and a measured target distance to Euclidean coordinates. The script traces a light ray through the system, where the light ray is represented as the solid arrows and is fully described by a starting point and direction vector. Two paths are shown leaving the ‘Position’ functions at the beginning of the figure; one is for 3D scanning and one is for cutting. Whether cutting or scanning is chosen determines which path is taken. The paths merge after the beam combiner, and the script continues to reflect the output from the combiner off of the mirrors in the galvanometer until it hits the target and outputs the intersection coordinates.

This same script can also be used to do the reverse and convert x, y, z coordinates to galvanometer rotations and distances. This reverse operation requires an iterative process where the galvanometer angles are first set to zero and the distance is set as the z-coordinate of the target. The script is run and x, y, z coordinates are received as output. A difference of angles is then calculated between the desired coordinates and the output coordinates. This angle difference is added to the commanded galvanometer angles and the script is re-run. This process is repeated until the output coordinates match the desired coordinates within a tolerance set by the user. By this methodology, the galvanometer angles can be calculated to precisely target any point in Euclidian coordinates within the surgical space.
FIGURE 2.9: Ray Tracing Script for Calibration
Success in surpassing the accuracy benchmark set by the human surgeon is contingent on careful selection and assembly of components as well as attentiveness to calibration. This chapter consists of three sections. First, a component-wise breakdown of the error introduced into the system is investigated. Second, the calibration methodology used to minimize assembly error is presented. The chapter concludes with an evaluation of the 3D-scanner accuracy and a characterization of the cutting laser.

3.1 Error Contribution by Individual Components

Each of the components used in the TumorCNC introduces some small amount of error to the accuracy of the full device in both cutting and 3D scanning. Quantifying this error is important to identifying and purchasing parts. Because each piece of hardware is itself a commercially-acquired device, the majority are accompanied by an accuracy metric as determined by the manufacturer. These accuracies are given as engineering tolerances where devices manufactured outside of these tolerances are removed from distribution.
during the quality control process. Within the academic environment however, researchers are generally more accustomed to working with statistical distributions where a device will have some probability of producing a range of values within a specified accuracy curve. A comparison of these two approaches is shown in Figure 3.1.

![Figure 3.1: Probability Density Functions for Accuracy](image)

Although the statistical accuracy distribution more likely represents the physical behavior of each component in the TumorCNC, the engineering tolerance distribution is used when calculating theoretical error. Using this distribution is a simpler but more conservative approach to accuracy estimation. Each of the errors quoted in the data sheet for each component are summed; the output is a single value for the full system accuracy, not a distribution. Although these error estimates will likely be much greater than in practice, within the surgical environment a single instance of broken tolerances can result in a fatal outcome for the patient.

### 3.1.1 Theoretical Error

A list of the potential error contributors is shown in Table 3.1 where the final column shows the theoretically-predicted error at the expected target range. For a perfectly calibrated system, it can be expected that the TumorCNC will exhibit a maximum lateral error of +/-117.5 m and a maximum depth error of +/-60.3 m when scanning.
Depending on the surface being scanned, the depth error is effected by the lateral error. If the surface is at a steep angle relative to the device, small errors in lateral positioning could change the depth quite severely. This error contribution is difficult to quantify, but it is likely that the maximum depth error will be larger than that quoted above depending on the quality and orientation of the scanned surface. On the other hand, cutting does not require the triangulation sensor so can expect a maximum error of +/- 52.5 m laterally. That said, if the cutting is based on information garnered from the 3D scanning, these values will be summed, indicating a maximum cutting error of +/-170 m plus any extra depth error as a result of lateral perturbations. In general, all of these errors lie well below the benchmark of 289 m set by the average surgeon. It should be noted that this is a breakdown by component – extra error will be imparted to the system through the calibration process to be discussed in the following section.

**TABLE 3.1: Theoretical System Accuracy – Breakdown by Component**

<table>
<thead>
<tr>
<th>Device</th>
<th>Error Form</th>
<th>Specifics</th>
<th>Angle [μRad]</th>
<th>Error [μm]</th>
</tr>
</thead>
<tbody>
<tr>
<td>DAQ: NI USB 6211 Output</td>
<td>Resolution</td>
<td>16 Bit, +/-10V Range</td>
<td>10.7</td>
<td>2.4</td>
</tr>
<tr>
<td></td>
<td>Absolute Error</td>
<td>Elec/Therm Noise</td>
<td>122.6</td>
<td>28.2</td>
</tr>
<tr>
<td>Distance Meter: MTI Microtrak 3</td>
<td>Dynamic Resolution</td>
<td>Radius of Triangulation Laser</td>
<td>-</td>
<td>65.0</td>
</tr>
<tr>
<td>Galvanometer: CTI 6246H</td>
<td>Resolution</td>
<td>Encoder Step Size</td>
<td>8</td>
<td>1.8</td>
</tr>
<tr>
<td></td>
<td>Zero Drift</td>
<td>15 μRad per °C</td>
<td>30</td>
<td>6.9</td>
</tr>
<tr>
<td></td>
<td>Scale Drift</td>
<td>50 PPM per °C</td>
<td>17.5</td>
<td>4.0</td>
</tr>
<tr>
<td>Galvo Driver: CTI MicroMax 671</td>
<td>Noise</td>
<td>Elec/Therm Noise</td>
<td>40</td>
<td>9.2</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Device</th>
<th>Specifics</th>
<th>Maximum Lateral Error when Scanning by Axis:</th>
<th>117.5</th>
</tr>
</thead>
<tbody>
<tr>
<td>Distance Meter: MTI MicroTrak 3</td>
<td>Processing Noise</td>
<td>Dynamic Resolution</td>
<td>32.5</td>
</tr>
<tr>
<td>DAQ: NI USB 6211 Input</td>
<td>Sensor and D-A</td>
<td>Depth Resolution</td>
<td>3.8</td>
</tr>
<tr>
<td></td>
<td>16 Bit, +/-10V Range</td>
<td>Resolution</td>
<td>10.7</td>
</tr>
<tr>
<td></td>
<td>Elec/Therm Noise</td>
<td>Absolute Error</td>
<td>93.9</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Device</th>
<th>Maximum Depth Error when Scanning:</th>
<th>60.3</th>
</tr>
</thead>
</table>
3.1.2 Electrical Noise and Isolation

The accuracy of each component given in Table 3.1 is only reliable if the electrical noise on the power and ground lines is kept below the levels specified by the manufacturer of each component. Consequently, it is important to test the signal lines for electrical noise to ensure the accuracy of the components will follow their theoretical prediction. The power supplies for each component were selected with equal or better ripple characteristics than recommended in their respective datasheets. Even so, with multiple supplies sharing a common ground, noise on a supply with a similar return path can affect the ground of another supply, potentially injecting noise into the signal lines. Unfortunately, all of the components communicate via analog voltage signals making them susceptible to electrical noise present within the system.

The signals most susceptible to noise are the outputs from the DAQ to the galvanometer and the input from the triangulation sensor to the DAQ. This noise is fairly straightforward to measure, and it is important to characterize it as it has a direct effect on the accuracy of the system. The measurements can be compared to the columns labeled as electrical and thermal noise in Table 3.1. The ripple present on the lines was measured by outputting a constant signal while the device was functioning and sampling it with an oscilloscope. A very basic oscilloscope was used with a single-ended, passive probe. This setup could have been improved by using an active, differential probe with an oscilloscope with higher resolution at low signal levels; however, this equipment was not readily available.

The first attempt at measuring ripple resulted in a large amount of ringing between 300 and 500kHz. After shortening the ground loop on the probe, ringing was significantly reduced but still present as shown in Figure 3.2. The other dominant frequency present
in the signal is at 100kHz; this is to be expected as that is the primary switching frequency of the power supplies. The $y$-axis labeled to the right of the figure translates the voltage units to what the corresponding steering error is at maximum range. The error due to ripple is less than 8 μm – well within the bounds set in Table 3.1 for the DAQ output. This outcome makes a more sensitive measurement with the equipment mentioned above unnecessary. However, it should be noted that the errors measured by this methodology do not include voltage offsets due to temperature or other factors. These errors would be exposed during the full system evaluation that follows.

**Figure 3.2: Noise on the DAQ Inputs and Outputs**

3.2 Calibration

Calibration of the TumorCNC is required to reduce any error present during assembly that would manifest itself as a misalignment of components. The error present from an
imperfect calibration is then summed with the theoretical errors presented in Table 3.1 to yield a full-system error estimate. The TumorCNC calibrates itself through use of its 3D scanner and the ray-tracing model described earlier. The goal of this calibration methodology is to achieve the maximum possible accuracy of the device by measuring and taking into account any offsets that could occur during assembly and alignment. This section is divided into two parts, where the first addresses calibration of the scanning portion of the device, and the second focuses on alignment of the cutting laser.

3.2.1 3D Scanner Calibration

The first step to calibrating the scanner is identifying possible sources of error in component positioning during assembly. As described previously, the coordinate system of the TumorCNC is defined with reference to the steering galvanometer. The coordinate system is shown in Figure 3.3, where the $x$-axis is aligned with the rotational axis of the $y$-galvanometer motor, and the $y$-$z$ plane contains the rotational axis of the $x$-galvanometer motor. Because the coordinate system is defined with reference to the galvanometer, it is impossible for any error to be introduced by its positioning.

Possible sources of assembly error are in the positioning of the beam combiner and the triangulation sensor. Each of these components can be oriented incorrectly with five degrees of freedom – two rotational degrees and three translational. When considering assembly, these 10 degrees of freedom are not independent. For example, changing the angle of the triangulation sensor can be expressed as a translation of the triangulation sensor combined with a change of angle for the beam combiner. Consequently, these 10 degrees of freedom have been distilled to 5 unique degrees of freedom: three translational degrees for the triangulation sensor and two rotational degrees for the beam combiner.
These degrees of freedom are depicted as orange arrows in Figure 3.3. The goal of this calibration is to scan a known geometry (in this case a flat plate at a known orientation) and predict the amount of error present for each of those five degrees of freedom.

The ray-tracing model developed for angular conversion (described in the previous chapter) is based on the as-designed geometry of the system where the components are assumed to be mounted in exactly the positions specified in the mount design. The calibration model uses this same ray-tracing model, but perturbs each of the five degrees of freedom within a specified amount of error. This method allows the ray-tracing model to simulate how the device would perform if the components were slightly misaligned. In this state of misalignment, the device simulates calibration data from a scan across a flat plate oriented perpendicular to the \( z \)-axis. A curve is then fit to this simulated data and compared to a curve-fit from calibration data taken by the device physically scanning across a flat plate and recording a distance at each point in the scan. The area between
these two curves is taken as a metric to how accurate each of the simulated perturbations are to the real assembly. This process repeats over a search region covering all possible perturbations with a given step size bounded by tolerances of the physical assembly process. The output metrics are compared and the perturbations corresponding to the smallest error between curves is taken as the current configuration of the physical system.

For further clarity, this process is depicted in Figure 3.4.

![Figure 3.4: 3D Scanner Calibration Workflow](image)

In practice, fitting a curve to this data will introduce some amount of error to the calibration process. One set of simulated data from a horizontal scan across the flat plate is shown in Figure 3.5 where the distances from the triangulation sensor to the plate are recorded on the right $y$-axis. Polynomials of different degree are fit to the data, and the distance between each data point and the curve is plotted on the left $y$-axis – this is the error introduced by the curve fitting process, or the quality of the fit. Although the data appears to be fit appropriately by a $2^{nd}$ order polynomial, it is shown by the bar graph that the error introduced by either a $2^{nd}$ or a $3^{rd}$ order polynomial is over 1 thousandth of an inch across parts of the plate. The $4^{th}$ order polynomial reduces this error by
approximately one order of magnitude as shown in the bottom graph in Figure 3.5. It makes intuitive sense that the 4\textsuperscript{th} order polynomial would significantly improve the fit as there are 5 degrees of freedom in the system.

![Figure 3.5: Comparison of Polynomial Fit for Simulated Calibration Data](image)

Both horizontal (\(x\) is changed while \(y\) is kept constant) and vertical scans (\(y\) is changed while \(x\) is kept constant) are used to calibrate the device with equal weighting. The physical scans occur on a flat plate manufactured to interface with the mount with close-fit dowel pins of high tolerance (shown in Figure 3.6). It is critical for the orientation of the flat plate and galvanometer to be as close as possible to the designed geometry because this is used as a truth source in calibration. It is estimated that this error will be
bounded at .002 inches due to manufacturing tolerances and the clearance for the pins. This error is then summed with the theoretical error because it cannot be calibrated out.

In order to improve the image quality of the triangulation spot, the planar calibration surface was media blasted with glass in order to roughen the previously-specular surface. Glass beads were chosen because they peen the surface, changing the shape of the material as opposed to removing material. Peening correlates to overall changes in dimension of much less than .001 inches. Beads were chosen with a mesh size between 170 and 350 (diameter between 44 and 88 microns) (McMaster Carr). This media size is approximately 3 times smaller than the spot size of the triangulation sensor, ensuring a large number of surfaces impacted by different beads will receive laser light from the triangulation emitter and reflect it back to the sensor. This peening will introduce a small

**Figure 3.6: Calibration Hardware**
amount of random noise into the scan, which not affect the measurement as a large amount of distance samples are be taken during each scan.

Figure 3.7: 3D Scanner Calibration Data and Error

The data and error from the calibration process is shown in Figure 3.7. Four vertical and four horizontal scans of ten thousand points each were recorded and averaged; each point from the averaged scans is represented by a black data marker. Fourth order polynomials were fit to this data as represented by the solid yellow line, and the simulation curve that produced the minimum distance between the calibration curves is plotted as a dashed yellow line. A detail view is shown identifying these different components. In similar form to Figure 3.5, the distances recorded are plotted on the right y-axis and the distance between the calibration curve and the best-fit simulation curve are plotted against
the left $y$-axis as bars. This distance is taken as a metric to the error introduced in calibration. When the dashed and solid curves are close to each other, it means the simulated calibration is providing very similar data to the physical calibration for that set of perturbations. This is easily shown by looking at the magnitude of the bars – which identify the distance between the curves. Consequently, when the bars are shorter, less calibration error is present.

As shown by the bars in Figure 3.7, the distance between the calibration and simulation curves is quite low. Especially towards the center of the plate where the surgery is most likely to be performed, the error between the two curves is significantly less than .001 inches. In order to honor the maximum theoretical error methodology, the maximum error in both scans will be used as the contribution to the error by the calibration process (.0018in – found at the positive end of the horizontal scan). Because the system relies on its own hardware to perform the calibration, error offsets present in those measurements will also contribute to the total theoretical error. These offsets are those errors present at the component level that do not have a zero mean; these include the DAQ’s absolute error and the galvanometer’s zero and drift errors. By summing the theoretical error presented in Table 3.1 with the maximum error in the calibration curve fit (.0018in/45.7μm), with the systemic hardware calibration error (.002in/50.8μm), and with the systemic component-level calibration error (.0015in/39.2μm), a number for the full-system error can be estimated. The sum of all of these errors yields a maximum lateral error of 253μm and a maximum depth error of 196μm; both of these figures lie a fair margin below the human benchmark of 289μm.

The perturbations associated with the simulation that best fits the calibration curves are given in Table 3.2 along with the full-system error estimate. Two sets of iterative
searches over the perturbation region were performed to reach these values, a coarse search and a refined search. In the coarse search, the translational search region was bounded to .02 inches from the designed position and the rotational search region was bounded to 3 degrees from the designed position. The step sizes during this search were .002 inches in translation and .125 degrees in rotation. After the first best-fit was identified, a refined search was performed over a constrained region with translational step sizes of .0001 inches and rotational step sizes of .005 degrees.

<table>
<thead>
<tr>
<th>Beam Combiner Rotation</th>
<th>Triangulation Sensor Translation</th>
<th>Full-System Error Estimate</th>
</tr>
</thead>
<tbody>
<tr>
<td>(\delta \phi)</td>
<td>(\delta x) (\delta y) (\delta z)</td>
<td>(Lateral) (Depth)</td>
</tr>
<tr>
<td>1.125(^\circ)</td>
<td>-0.75(^\circ) 0.0002 in 0.0044 in -0.0017 in</td>
<td>0.0100 in 0.0077 in</td>
</tr>
</tbody>
</table>

### 3.2.2 Cutting Laser Calibration

Calibration of the cutting laser is performed by lasing holes in surfaces, measuring the holes with the 3D scanner, and using the position of these holes to triangulate the position of the cutting laser. Similar to calibrating the 3D scanner, the as-designed geometry is perturbed until the simulated holes appear as close as possible to the physical holes. The summed norm between the simulated holes and the physical holes is taken as a metric to the success of the calibration process. White computer paper was used as a substrate for hole creation because it yields reliable measurements with the distance sensor. The position and orientation of the paper relative to the device is not critical because these factors are measured by the 3D scanning process after lasing.

After a hole is created in the paper, a small patch around the hole is scanned. A plane is fit to the outer 10% of this scan (depicted as the orange data points in Figure
3.8); all of the scanned points are then projected onto this plane. The user is then prompted to identify a point within the bounds of the hole. The identified point is taken as the origin of the plane, and all points are converted into polar coordinates. The plane is then divided into a number of pie slices radiating out from the origin. Within each pie slice, the nearest point is identified. The average of these points is taken as the center of the hole.

This process is repeated with the origin being defined as the center of the hole at every iteration until its position change at each iteration drops below .00001 inches. Without varying the number of pie slices, this method of center identification is repeatable up to .0001 inches no matter where in the hole the origin is first identified. The number of pie slices is not critical, as long as there are enough to identify the majority of the points on the edge of the hole without selecting many points beyond the edge of the hole. For the set of scans used for this calibration, the number of slices was taken to be

![Figure 3.8: Identifying Hole Centers from Projected Scans](image)
approximately 40. Figure 3.8 shows the points of a scanned hole projected onto a plane and identifies those points used to specify the center of the hole.

The position of each of the holes as well as the galvanometer mirror positions are fed into the script used to calibrate the cutting laser by perturbing its position. With four degrees of freedom, at least four holes are required to uniquely identify the position of the cutting laser. These holes are ideally located at much different x, y, z locations to increase the triangulation accuracy. More holes may be measured to improve the accuracy.

For the current calibration, 9 holes were shot and scanned across the full range of positions available to the galvanometer. Although points were chosen near each of the four outer corners of the field-of-view, multiple points were selected towards the center because calibration accuracy is critical where cutting is most likely to occur. The outcome of the calibration is shown in Figure 3.9 where the hole positions are projected onto the x-y plane and plotted against the right axis. The bars show the distance between each of the physical holes and the simulated holes with the best-fit calibration perturbations (as plotted against
the left axis). It should be noted that the largest error is approximately .005in/127µm – less than half of the required precision of 289µm.

The perturbations corresponding to the calibration above are given in Table 3.3. Notice that the translation values are significantly larger than the 3D scanner calibration. This is due to the cutting laser alignment occurring by eye as opposed to automatic alignment through precision-machined components.

<table>
<thead>
<tr>
<th>TABLE 3.3: Best-Fit Perturbation Values for Cutting Laser Calibration</th>
</tr>
</thead>
<tbody>
<tr>
<td>Rotation</td>
</tr>
<tr>
<td>δθ</td>
</tr>
<tr>
<td>1.8°</td>
</tr>
</tbody>
</table>

3.3 3D Scanner Accuracy Testing

After performing the calibration methodology outlined previously for the 3D scanner, a second series of tests was required to fully characterize the accuracy of the device. These tests are performed by placing objects of known dimensions in the path of the 3D scanner, performing scans, and comparing to truth data.

The calibration of the cutting laser immediately unveiled a unique issue with the 3D scanner. The hole shown in Figure 3.8 is a projection of the 3D data, while Figure 3.10 shows a 3D view of that same scan with an inset depicting a magnified view of the hole shot through relatively planar piece of white paper. Clearly the 3D data does not match the physical geometry of the hole. With some experimentation, it was found that the scanner introduces artifacts for features perpendicular to the x-axis while features perpendicular to the y-axis are normal (i.e. a vertical edge vs. a horizontal edge). This behavior can be credited to the parallel orientation of the triangulation sensor receive
array to the $x$-axis. Implications are discussed in the following chapter, but this realization helped identify what geometries should be evaluated to quantify the accuracy.

The primary test piece was a slot with a width of .031 inches and a depth of .02 inches machined from a small piece of aluminum. This slot was scanned by a white-light interferometer (Zygo Corp., Middlefield, CT) to get an accurate depth measurement. The slot was then scanned by the 3D scanner in both the horizontal and vertical position. The scans are shown in Figure 3.11 and Figure 3.12, where the upper plot in each depicts the 3-dimensional view of the gathered point cloud, and the lower plot in each is a projection of that point cloud onto a plane perpendicular to the length of the slot. The previously-designed accuracy benchmark of 289µm is included for reference.

**Figure 3.10: Triangulation Artifacts at Sharp Edges**

![Triangulation Artifacts at Sharp Edges](image)
As indicated on the lower plot of each of these figures, the small black points are those sampled by the 3D scanner, the blue points are a 100-point moving average across those points (100 points were scanned in depth), the yellow shaded area indicates one standard deviation on either side of the average, and the thick, black data clusters indicate the truth value as measured by the white-light interferometer. The data was gathered by placing the slot near the middle of the triangulation sensor range. The mirrors scanned
across a 100x100-point grid with equal steps in x and in y; the distance was sampled at each point. The distances and angles were then converted to x, y, z coordinates via the ray-tracing model presented previously. Because the slot was machined in aluminum, the surface was powdered with an aerosol spray (GoldBond® Chattem Inc., Chattanooga, TN) to prevent specular reflections during scanning.

**Figure 3.12:** Vertical Accuracy Testing – 3D and Projected Views
Almost every data point for the horizontal slot lies within the accuracy benchmark of the truth data. The first standard deviation is well within the benchmark and the average value settles very close to the truth value. Alternatively, the vertical slot shows large deviations from the truth value. For the most part, the average and standard deviation also lie within the accuracy benchmark, but the geometry does not resemble the vertical slot nearly as well as the horizontal test.

The variance of the distance data appears higher than predicted by the values given in Table 3.1, even with the consideration that each of the data points presented in the projected views was gathered at a different position on the powdered surface – potentially adding to the variance. To investigate the variance introduced by the triangulation sensor, follow-up testing was performed where the distance sensor was repeatedly sampled at a single galvanometer position.

**Figure 3.13: Triangulation Sensor Repeatability Probability Density Function**

Two tests measured points on the powdered-slot surface, and two measured a semi-opaque sheet of white acrylic (results presented in Figure 3.13). The acrylic was tested to
investigate the variance added by a surface into which the triangulation laser could partially penetrate. For the two scans on each surface, the galvanometer mirrors were kept stationary for the first, while for the second, the galvanometer mirrors were moved to a different position and returned to record the distance at the same point. These ‘semi-stationary’ samples were taken to mimic galvanometer movement during a procedure and investigate any contribution of variance due to the variance in mirror positioning as well as triangulation. In order to increase sensitivity to variance in the galvanometer repeatability test, the surfaces were oriented at a 45° angle to the output of the mirrors.

The results show little change between the stationary and ‘semi-stationary’ tests for both surfaces; this is indicative of high galvanometer repeatability – at least high enough to be lost in the noise of the triangulation-sensor. All of the scans were highly Gaussian, and consequently, the raw data was not plotted except in the case of the ‘example data spread’. The acrylic scans did show a slight tail to one side of the Gaussian which is a result of the laser penetrating further into the material. The results from the acrylic scan spread a much wider range than those of the powdered surface. This occurs because the laser spot is ‘stretched’ as it penetrates the material. Overall, the surface properties effect the measurement dramatically. A coarse surface with diffuse reflections will return a spread with a standard deviation of approximately half the desired benchmark of the system; a semi-translucent surface, however, will provide a much wider range of distances sensed.

3.4 Cutting Laser Characterization

Describing a few specific characteristics of the laser-beam incident on the surgical target is critical to understanding and predicting the laser-tissue interaction. In its current state,
the computer has the ability to change the power and the pulse-width of the laser through user commands while the spot size and focal point are changed by physically replacing parts. A mapping between the control PWM signal and the output power is required so that the laser can supply the appropriate amount of energy as desired by the planning algorithm (currently under development).

Characterization of the PWM-to-output behavior was performed by varying the duty cycle while measuring the energy incident on a thermopile power meter (Newport Corporation, Irvine, CA). Because a thermopile is generally used for measuring long laser pulses, 3-second pulses were used. During the first 1.5 seconds of each pulse, the thermal gradient was established without sampling, then 4 samples were recorded and averaged over the following 1 second interval. Each pulse was followed by a 3 second relaxation time where the thermal energy was allowed to dissipate. A single pulse was measured at every 5% increase in duty cycle up to 100%. Each of these scans was repeated 50 times to create the mapping shown in the upper plot of Figure 3.14. The black points represent measured data and the shaded area represents one standard deviation on either side of the mean.

The standard deviation was larger than expected for the laser output, and further investigation was required. Consequently, a continuous wave was measured at 40% duty cycle as shown in the bottom plot of Figure 3.14. Ideally, this would output a constant power, but the output fluctuates in a strange waveform at an amplitude no greater than ±8%. After checking the stability of the power supply, the PWM accuracy, and comparing manuals between the laser and the power meter, this behavior was credited to the laser with the manual quoting an accuracy of ±10% [38]. This behavior will likely cause some difficulty in planning, and will be discussed further in the following chapter.
Independent from the PWM-to-power relationship is the output beam profile of the laser. This is generally described on a cross section through the axis of the beam as a variation in intensity as a function of distance from the axis. As discussed in the previous chapter, a spatial filter became part of the design in order to ensure a Gaussian profile at the output of the final focusing lens. Unfortunately, during alignment the pinhole used to perform this filtering was perforated by the laser. Occurring at an untimely moment, the experiment proceeded without the filter and the pinhole was found to be unnecessary. This does not change the design in any way, as the same lens-replacement methodology described earlier can be used to alter spot size. Without the pinhole, the laser can also be run with a single focusing lens to achieve the medium spot diameter described in Table 2.1, and this is how it was configured for the following experiments.
There are multiple methods of measuring spot size; the most accurate require
delicate imaging equipment. The economical method utilized for this laser is slightly
more involved. The method requires a thermopile power meter to be partially occluded by
an object with a sharp edge – in this case, a razor blade. The razor blade is slowly advanced
across the face of the power meter, and at every step, the power is measured. At each step
across the meter, the power drops until it has reached zero, thereby outlining the power
profile of the beam. In the case of the TumorCNC, the scanning galvanometer allows the
razor blade to remain stationary while the beam is scanned across the face of the power
meter. This process is demonstrated in the upper rendering of Figure 3.15 where a single
sample is shown with the razor blade partially occluding the power meter from the laser
beam. Because the majority of the power meter is exposed to the laser energy, this
corresponds to a higher value of measured power as identified by one of the points in the
middle plot.

The spot size was measured at two different z-locations ½-inch apart. The yellow
points correspond to the further measurement while the orange points are from the close
measurement. After taking 10 full scans across the razor blade, the mean and standard
deviation was calculated at each measurement point. A Gaussian error function was fit to
points representing one standard deviation on either side of the mean. The lowest R² value
for the fit was .9995, signifying an excellent fit, thereby indicating that the beam profile
is indeed Gaussian and that a spatial filter is likely unnecessary. The constants found
during the fitting of this curve were then used to plot the corresponding beam profile in
the lowest plot, where again, the shaded region accounts for one standard deviation from
the mean. The 1/ε² beam diameter is also plotted for each of the two spot sizes measured.
The points were measured at a ½-inch distance for a variety of reasons including ½-inch being the extent of throw for the translation stage used, the energy density of the beam needed to be below the damage threshold of the meter, and it is necessary for this measurement to occur far from the beam waist so the mode of the laser does not alter the results. These measurements are used to provide a rough estimation of the divergence of
the beam, and the location of the focal point with reference to TumorCNC-centric coordinate system as shown in Figure 3.16. The beam is shown as the blue solid lines, while the beam waist diameter predicted in Table 2.1 is shown at the focal point by the dotted blue lines.

![Figure 3.16: Measured Focal Point Location](image)

**Figure 3.16: Measured Focal Point Location**
There are two primary objectives of this work. The first is to present and justify the important design choices made towards the development of the first-prototype TumorCNC. The second is to introduce a method for calibration and evaluate the accuracy of the device. This document addressed both of these objectives by the order in which they were undertaken. In this chapter they will be introduced in reverse order; a discussion of the performance of the current device will be succeeded by and inform recommendations for future development.

4.1 Performance and Design Recommendations

As outlined in the first chapter, time and accuracy are the primary metrics by which the first prototype of the TumorCNC can be evaluated. As a hardware-based thesis, the majority of the effort was dedicated towards ensuring the TumorCNC can meet the 289μm accuracy benchmark. With a direct impact on operating cost, however, time has an almost equal impact on the success of the system.
4.1.1 Operating Time

As it currently stands, there are a variety of roadblocks preventing the first-prototype TumorCNC from executing a simulated operation, and one of these is operating time. Although the first prototype is not yet ready to be tested for the time required to perform a resection, the hardware was chosen with the requirement that an automated resection be performed in less time than a standard resection (as outlined in Chapter 1).

Being able to compare the performance against standard resection times requires first identifying a typical operation against which the device will be benchmarked. Our group has targeted the removal of a superficial, small, non-infiltrative, and well-circumscribed tumor (clinically represented by a cortically-based brain metastasis) as the initial application. The amount of time required to remove a tumor such as this varies considerably, but according to collaborating neurosurgeons, a characteristic timeframe would be approximately 2-3 hours per tumor. These numbers will be more carefully defined during a complete market analysis to take place at a different date.

The time required for the TumorCNC to remove a tumor is dependent on both the time spent scanning and the time required for the cutting laser to remove all of the tissue. In its current embodiment, the TumorCNC takes .025 seconds to move to a point and measure the distance to that point by the discrete method of control described in the design section. For a $3\text{cm} \times 3\text{cm}$ tumor and 250$\mu$m resolution, it would take 6 minutes to complete a full scan of 14400 samples. Assuming the tumor is cubic, 250$\mu$m of tumor is removed at each cut, and a full 3D scan is performed between every cut, it would take 12 hours just to perform the 3D scanning with the current scanning control software.

With effort dedicated towards configuring triggers on the DAQ, the discrete scanning software can be altered to perform continuous sampling (each scan is loaded as
a task on the DAQ instead of each sample, significantly reducing time spent initializing
tasks). This change could increase the scan speed of the system to its hardware limit of
10kHz (a function of the maximum sampling rate of the triangulation sensor at 20kHz).
This increase in speed would reduce the time spent scanning during the entire surgery by
approximately 250 times, thereby reducing the time spent scanning from 12 hours to just
under 3 minutes. With such a small amount of time spent scanning, it may be possible to
execute multiple scans, or perform scans of higher resolution for better accuracy.

The time spent cutting is a function of the amount of power the laser can output
and the ablation efficiency of the tissue. Estimating the tumor to have approximately the
same density as water, it would weigh approximately 27 grams for the geometry used in
the previous calculation. It has been shown that a 112W CO$_2$ laser with a pulse width of
160µs removes approximately 4 micrograms of tissue per pulse [49]. Although the ablation
threshold does not necessarily remain constant between pulsed operation and continuous
operation, a rough approximation yields 40 seconds to remove 1 gram of tissue under
continuous operation with those laser specifications, or approximately 18 minutes to
remove the cubic, 3cm tumor under continuous operation.

From Figure 3.14, it was shown that the laser installed on the TumorCNC can
operate at approximately 11W; all other laser specifications are very similar to those given
in [49]. This indicates that a laser with higher power will likely be required before the
TumorCNC can be used to perform a large-scale resection because the time required to
perform a full operation on a 3cm tumor would be much too large. The use of other lasers
with the TumorCNC is discussed in Section 4.2: Future Development.

To summarize, prior to performing a large-scale simulated resection, the
TumorCNC would require software to run scans in a continuous mode, and it would require
a laser with approximately 10-times the power of the current laser. Neither of these changes would require a significant time investment by the team, but depending on the laser selected, this choice could incur significant cost. With these changes implemented, the time required by the TumorCNC to resect a 3cm tumor is approximately 20 minutes – significantly lower than the current estimated operating time.

4.1.2 Scanning Accuracy

Aside from operating time, the other metric by which the first-prototype TumorCNC is evaluated is precision. Reaching the precision benchmark set by the human counterpart of the TumorCNC was the primary focus of this thesis. Figure 3.11 shows success in reaching this precision for one geometry while Figure 3.12 shows marginal results to be construed as failure for the same geometry rotated at 90°. For sharp edges with some component oriented vertically along the \( y \)-axis, the triangulation sensor introduces artifacts which do not represent the physical object being scanned.

The triangulation sensor functions by emitting a laser and looking at the return image on an array of CCD pixels located off to the side of the laser. When the return image falls on one side of the array, the sensor registers a far measurement while a return image on the other side of the array registers a close measurement. Because the emitted laser has a finite diameter, the return image is thresholded and the centroid is taken as the distance. Unfortunately, this process of thresholding and taking the centroid is what causes artifacts with edges perpendicular to the \( x \)-axis.

Figure 4.1 shows a detail view of the triangulation sensor attempting to measure along an edge. Diagram A shows an example of the sensor registering the correct distance. The full Gaussian laser beam is incident on the receive sensor which correspondingly
outputs the correct distance after taking the centroid of the illuminated pixels. Unfortunately, when one side of the Gaussian beam is either attenuated or scattered in a different direction, the receive sensor centroids only part of the beam. This causes the sensor to register an object as being closer than reality (as shown in diagram B) or as being farther than reality (as shown in diagram C).

![Diagram of sensor readings](image)

**Figure 4.1: Edges Cause Incorrect Triangulation-Sensor Readings**

A number of objects with different geometries were scanned in order to characterize the performance of the triangulation sensor for vertically-oriented edges. The most revealing scans are those where gauge blocks were placed directly next to each other and an edge was created. The response is different based on whether a region shaded from the line-of-sight of the sensor is created. Figure 4.2 and Figure 4.3 show the different responses to shaded and unshaded regions. The unshaded geometry creates a more symmetric anomaly while the shaded geometry creates an asymmetric artifact.
**Figure 4.2:** Vertical Edge Scan with Shaded Region

**Figure 4.3:** Vertical Edge Scan without Shaded Region
A variety of methods could be used to reduce the magnitude of artifacts present along vertical edges. The simplest approach is first to remove unphysical points from the scan. If two points are found to lie behind each other from the point-of-view of the sensor, they can be removed. It may also be possible to filter for and remove waveforms similar to those shown for the vertical edges. These methods should be able to remove the most egregious artifacts from the scan, likely bringing the accuracy below the benchmark accuracy of 289µm.

Outside of signal processing, a variety of methods for artifact reduction can be attempted in hardware. The magnitude of the artifacts is a function of the diameter of the laser emitted by the triangulation sensor. Consequently, if the diameter of the laser can be reduced or the threshold of the sensor can be raised prior to taking the centroid, the size of the artifacts can be significantly reduced. For example, if the laser is small enough to fit within a single pixel of the receive sensor, the artifacts would not exist. Although it is not possible to change the sensor threshold without modifying the device, it may be possible to attenuate the emitted beam with a neutral density filter, essentially bringing the threshold closer to the maximum intensity of the beam. It may also be possible to install optics to provide a tighter focus for the emitted beam. In both cases, care must be taken to avoid changing the beam path and altering the measurements although it is likely any errors can be calibrated out by the same methodology already presented.

At this point it is fair to say that the desired accuracy of the 3D scanner was achieved by a large margin for those geometries in which artifacts are not present. The next objective will be to implement methods of alleviating artifacts such that the benchmark for accuracy is reached 100% of the time.
4.1.3 Cutting-Laser Accuracy

The cutting laser receives its direction commands based on data measured from the 3D scanner. Consequently, its accuracy can be represented as the accuracy of the 3D scanner plus any extra error added during calibration of the cutting laser. The calibration introduces .005in of error which means that points identified by the 3D scanner can be shot by the cutting laser within approximately .005in. At this point, the most influential method of improving accuracy is to improve the repeatability of the 3D scanner.

The largest difficulty currently present with this laser is its inability to hold constant power. The output power fluctuates by approximately 8%, posing a relatively large challenge to planning. Because the fluctuations are a function of the total output power, in sensitive areas the output power must be reduced to avoid overshooting into critical tissue. This approach will ease the planning of each cut, but will lengthen the surgical time. The fluctuations in power are a physical phenomenon that originates from heating of the tube containing CO$_2$ gas. As the tube heats and cools during operation, it’s overall length changes and effects the amount of energy that is emitted. The only way to reduce this would be to attempt to keep the laser in thermal equilibrium which could be attempted through running the laser for an extended amount of time prior to an operation (unlikely), or by heating the laser prior to operation (the impact will have to be investigated).

It is likely that a new laser with more power is required for large resections. The current laser was purchased as a low-power research laser for demonstrating proof-of-concept and performing research related to laser-tissue interaction. Simply replacing this laser with a more accurate model would solve this problem.
4.2 Future Development

This thesis outlines the first attempt at creating an automated tumor ablation device. Through demonstrating improved precision and reduced operating times from the current neurosurgical methodology, semi-automated, stereotactic laser resection stands to change the face of neurosurgical tumor removal. The first-prototype TumorCNC is the first step towards realizing this goal. To fully achieve the desired speed and accuracy outlined in Chapter 1, a large amount of development still remains. These efforts can be categorized into three primary thrusts: imaging, registration, and cutting.

4.2.1 Imaging

Three-dimensional scanning was chosen as the primary method of imaging on the first prototype for a variety of reasons including cost, simplicity, and availability. A variety of other sensing methods can potentially be included on the TumorCNC platform to either improve accuracy, or allow for some level of diagnostic capability.

Arguably, the simplest sensor to include is a visible-light camera. At the bare minimum, this camera would be required to provide a video-feed to the surgeon as a method of increasing situational awareness. This camera could also be used to increase the accuracy of the current 3D scanner through providing a second, independent level of triangulation and a method to identify edges within the image. The artifacts introduced in the current surface scans are a result of vertical edges present on the scanned object. A visible-light camera could be used to identify vertical edges within the scanned region and flag these as areas in which artifacts could be present. A higher threshold of filtering could then be applied to these specific areas of the scan. The camera could also be used to perform triangulation against the same laser emitted by the current triangulation sensor.
With careful calibration of the camera position, the camera could use the reflection and pointing vector of the emitted laser to triangulate the distance to a point. By placing the camera vertical to the current setup, the artifacts introduced through the same physics would appear for horizontal edges as opposed to vertical edges. By fusing the triangulation data from the two sensors, those areas presenting discrepancies could either be filtered or simply removed. With two sensors performing triangulation on the same object, the amount of artifacts could be reduced significantly.

Other imaging techniques may also lend themselves to improving accuracy of the system. Specifically, optical coherence tomography (OCT), primarily used as a penetrating form of imaging for ocular surgery, could lend itself well to surface mapping in this scenario. This idea has not yet been fully explored.

Other imaging techniques of interest to the project lie in the area of combined resection and diagnostics. As demonstrated in [31], fluorescence could potentially be used to identify those parts of the anatomy that are cancerous as opposed to those that are not. After exciting the entire surgical region with the fluorescent activation wavelength, those areas that re-emit would be specifically targeted by the path planning algorithm as tumor cells. A unique approach at tumor-cell targeting has been demonstrated previously at Duke University and will be further investigated for applications with the TumorCNC [50].

4.2.2 Registration

With an imperfect sensor stream injected into the system, it is important to be able to process and use this information to correctly indicate those areas of anatomy that had been previously labeled by the surgeon as cancerous. The primary challenge is registering a 3D model of the surface with a 3D model of the MRI. There are many ways of solving
this problem, and none have yet been attempted within our group. Our team has identified potential collaborators in the department of mathematics at Duke University. Their work in aligning CT-acquired surface scans of lemur teeth could potentially lend itself very well to the alignment of the point cloud generated by the TumorCNC [51].

4.2.3 Cutting

The final piece of the puzzle is the question of the cutting laser. Although the CO$_2$ laser has been thoroughly researched for laser-tissue ablation applications, many questions remain unanswered. The next step in this work is to select and implement a laser-tissue interaction model that can be used to predict, based on specific laser and tissue characteristics, what the ablation efficiency of the tissue is. Without this prediction, the ability to plan a cutting path is left to guessing and measuring previous cuts. This model will be developed and tested for the current laser, but it is very possible that the choice of laser changes in future iterations of the device. It is important to make the model flexible and nonspecific, such that if a new laser is selected it can be easily integrated.

The choice to change laser would likely be motivated by a desire for cleaner ablation characterized by less thermal spread. Lasers with extremely short pulses have been shown to perform very clean ablation, and discussions are ongoing with Synaptive Medical to onboard their very-capable, picosecond infrared laser (PiRL) [52]. Unfortunately, with less thermal spread, the coagulation ability of the laser is also significantly reduced. Consequently, it may also be necessary to include a second, co-aligned laser on the TumorCNC with a wavelength tuned for coagulation. As with the other three research fronts, this is an ongoing effort.
Bibliography


